THREE-DIMENSIONAL NUMERICAL SIMULATION OF HUMAN KNEE JOINT MECHANICS

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ABSTRACT

Objective: The knee joint is the part of our structure upon which most mechanical demands are placed and a large number of lesions are associated to it. These factors motivated the construction of a three-dimensional model of the human knee joint in order to simulate joint kinematics and obtain the mechanical demands on the main ligaments during knee flexion movements. Methods: The finite elements method was used to build a three-dimensional, biomechanical model of the knee joint. In this model with six degrees of freedom, the flexion/extension movement is applied, while the other five degrees of freedom are governed by the interactions between joint components. Results: Data was collected on the movements, on the internal/external and adduction/ abduction rotations, on the anterior/posterior, lateral/medial and upper/lower translations, and on the forces acting upon the four main joint ligaments, during a wide flexion/extension movement. These values were qualitatively compared with comparable values available in the literature. Conclusions: It was observed through an analysis of the results that several kinematic aspects are satisfactorily reproduced. The initial pre-load of the ligaments and the positioning of the ligament insertions in the model were shown to be relevant variables in the results.

Keywords: *Biomechanics. Statistical analysis. Finite element analysis. Knee joint.*

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INTRODUCTION

Anatomy was traditionally based on experiments in animals and human beings aiming to achieve a better understanding of the biological structures. Mastering this science allows us to enhance surgical procedures' effectiveness and to study the development of new therapy methods for musculoskeletal system pathologies.

The knee is the greatest and most demanded joint in human body, being composed by femur, tibia, fibula and patella bones, attached to support and stabilization structures such as ligaments, joint capsule, menisci and muscles.

Due to the high mechanical demand to which it is submitted in its support function, a large number of injuries are associated to it, such as total and partial ligament ruptures, meniscal fissures and injuries, bone fractures, osteochondral injuries, and others.

In terms of kinematics, human knee is a hinged system with 6 DOF – degrees-of-freedom, enabling combined movements not depending on rotation and translation, with flexion/ extension being the key movement (rotation around axis x). The remaining degrees of freedom are the upper/lower translations (translation along axis z), medial/ lateral translation (translation along axis y) and internal/ external rotations (rotations around axis z) as abduction/ adduction (rotation around axis y). Figure 1 shows an illustration of these degrees of freedom.

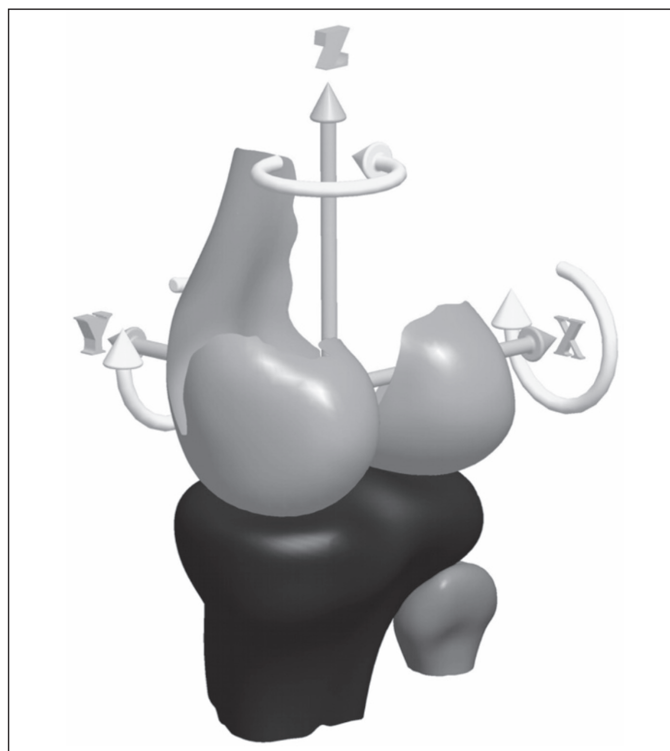

Figure 1 – *Knee movements.*

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The biomechanical function of the knee joint is ruled by the complex interaction between its components: the patella, the distal femoral portion and the proximal tibial portion, cruciate ligaments, collateral ligaments, synovial capsule, joint cartilage and menisci, as well as muscles. Due to a complex interaction between its components, any damage to these can lead to an unbalanced natural biomechanics of the knee and promote impairment of the whole joint system.

Ligament injuries are common events in adults, especially anterior cruciate ligament injuries, which basically occur during sport activities and as a result of car accidents. Studies suggest an incidence of 0.03% of ACL injuries each year among U.S. population. In these cases, although good clinical results are achieved soon after ligament reconstruction surgery, 20-25% of the individuals show some kind of postoperative complication, including osteoarthritis and instability, and these phenomena can progressively cause damages to other knee structures. It is worthy to mention, however, that ligament injuries do not depend on overloads only. Previously conducted studies indicate that hormonal variations impose a significant influence on the ligament rupture mechanism.[1]

The anterior cruciate ligament (ACL) is attached to the femur on the posterior intercondylar area and on the medial surface of the lateral femoral condyle, being fixated to the tibia on the anterior intercondylar region. The posterior cruciate ligament is attached to the femur on the posterior intercondylar area on the lateral surface of the medial femoral condyle and on the tibia at the posterior intercondylar region, with these positions determining its functions. At the intermediate position of the flexion/ extension movement, ligaments help tibia and femur to attach, and, at extreme flexion/ extension positions, they act by limiting anteroposterior displacements, with the Posterior Cruciate Ligament acting by limiting the posterior drawer movement (tibial backwards translation) and the Anterior Cruciate Ligament on the anterior drawer movement.

ACL reconstruction is a complex procedure, with several surgical variables involved affecting the ability of a graft replacing the ligament to restore knee joint function. Some of these variables are more frequently studied, including the positioning of graft fixation to the bones, the fixation method, the graft material and the pre-tension applied to the graft at the moment of fixating it. Literature, however, usually presents contradictive conclusions. Some authors advocate that the pre-tension applied to the ACL graft at the moment of reconstruction should be light in order to minimize the risks of graft rupture during its use, as well as to reduce contact tensions on knee's joint surfaces. Other studies advocate that a strong tension applied to the ACL graft at the moment of reconstruction would be beneficial for joint stability. Thus, despite of the large number of in vitro studies showing that the pre-tension applied to the anterior cruciate ligament (ACL) graft at the moment of fixation tends to affect the normal stability of the knee joint, most studies report almost no difference in the long term, suggesting the occurrence of a balance of tensions on grafts after some time postoperatively, associated to the new ligament remodeling on the graft and the necrosis process occurring on the implanted graft.[2,3]

In this sense, an enhanced knowledge of the kinematics of human knee is very important for studying treatments for this joint's diseases.

OBJECTIVES

This study has as an objective to build a three-dimensional model for a normal human knee joint that could allow us to reproduce its kinematics, intending to simulate the mechanical forces imposed to ligament insertions during knee's flexion movements.

METHOD

Geometrical Model

Geometrical models of anatomical pieces are difficult to obtain and manipulate, especially due to their irregular surfaces. The three main methods employed for characterizing the geometry of anatomical pieces were the following: measurements by coordinates of cadaver's anatomical pieces, laser scanning of cadaver's anatomical pieces, and geometry measurements from magnetic resonance or in vivo computed tomography.

The geometrical model employed in this study is constituted of the femur, tibia and fibula, obtained from the Biomechanics European Laboratory, as shown on Figure 2.

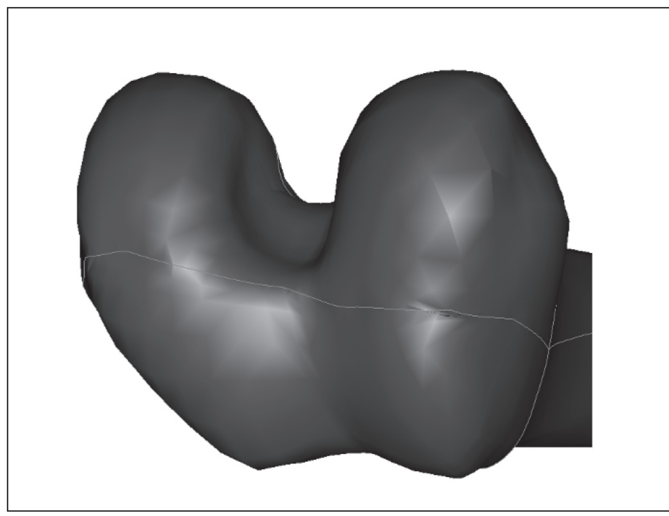

Figure 2 – *Geometrical model of the femur by the Biomechanics European Laboratory.*

For joint surfaces to be used in simulation processes, it is convenient that these are continuous and smooth. This enables the generation of quality superficial elements meshworks, slight early distortion of the elements, and easy treatment of contact conditions between the surfaces. In order to adjust the original models to these conditions, the retro-engineering software GEOMAGIC STUDIO® was employed, which imports geometrical files, transforming them into a dotted scatter. Thereby, it provides a homogenous distribution of this scatter, making a triangulation between the dots and, from which, generating NURBS surface patches forming a new surface, now seamlessly between adjacent NURBS patches. The resulting geometrical model of the femur from these operations is represented on Figure 3.

Despite of the importance of menisci for knee stability, we couldn't provide a model for it because no consistent geometrical description was available for the bone set geometry. This simplification was similar as the one adopted by Blankevoort and Huiskes[4] and by Song et al.[5]

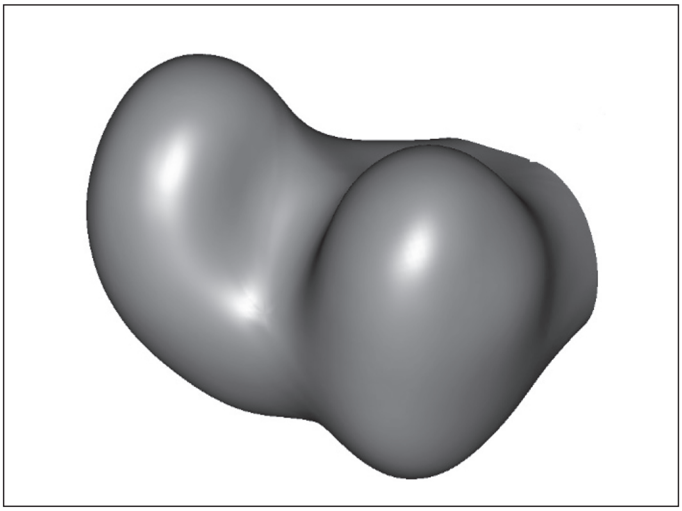

Figure 3 – *Geometrical model of the femur after treatment with the Geomagic Studio software.*

NUMERIC MODEL

For generating a numeric model, the geometrical model was imported by the ICEM® software, where ligaments were built, meshworks were generated on bones geometry, and contact pairs, outline conditions, properties, material models and loads were determined. The mechanical analysis employed the Finite Elements Method, with the commercial code ANSYS® used as a solver.
Since the bones forming the knee experience much less deformation than ligaments and for not being the object of our study, these were regarded as virtually non-deformable compared to ligaments, and casted with peel elements on the joint surface, with stronger stiffness in order to preserve its geometry. Therefore, the number of degrees of freedom is significantly reduced in comparison to a solid model.
Ligaments and tendons are dense connective tissues, constituted of few cells called fibroblasts, soaked into a large intercellular matrix, corresponding to about 80% of the total ligament volume. This matrix is formed by approximately 70% water and 30% solid material. Type-I, III and V collagen, and glycoproteins and elastin represent the largest portion of this solid material.
This structure of soft connective tissues gives rise to a complex interaction between fibroblasts and the intercellular matrix, promoting a viscoelastic property to the mechanical behavior of ligaments. In traction assays with ligaments, the tension-deformation curve format is dependent on the deformation rate in which the test is performed. The effects of the deformation rate on the tension-deformation curve have been extensively studied, on anterior cruciate ligaments, tendons, wrist ligaments, periodontal ligament, among others. In the present study, the studied movements are admittedly performed at low speed. In this case, viscous effects are negligible, and purely elastic models become appropriate.
Three-dimensional models of ligaments require the use of anisotropic constitutive or cross-sectionally isotropic relationships.[6,7] Uniaxial elements are particularly appropriate to simulate these components, due to its mechanical properties aligned with its cross-sectional geometry strongly below its length. In this case, only a one-dimensional constitutive relationship is required. The data concerning this relationship between tension and deformation of the ligaments were obtained by Mesfar and Shirazi-Adl[8] from which a principle for ligaments' behavior was built. The one-dimensional tension-deformation curve used in this study is shown on Figure 4, which is applied to all ligaments.

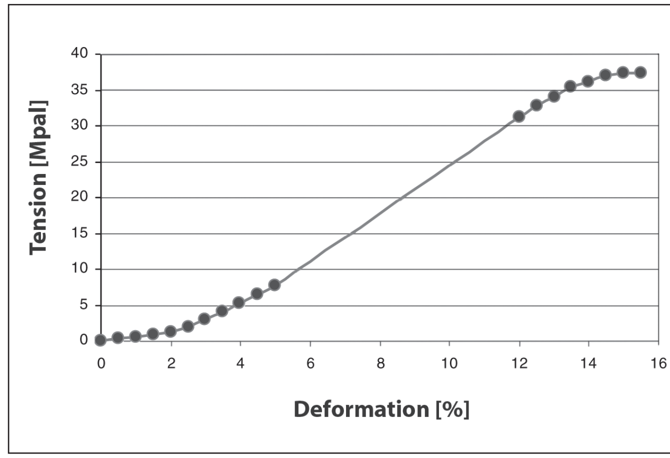

Figure 4 – *Tension-deformation curve applied to ligaments.*

Ligaments almost can't resist to compressive efforts or flexion/ torsion efforts, thus constituting a sort of elastic rope. However, they have tonus, also called initial or pre-load tension, which provides them a pre-traction condition even in muscular rest situations.
In order to promote pre-tension on ligaments, a numeric resource was employed which consisted of including a thermal expansion coefficient on ligaments. For each one, a thermal load required for generating an initial deformity to promote a desired tension was applied.
Thus, ligaments are represented by uniaxial finite elements defined by 2 knots (3 GL/ knot) responding to axial efforts only. This model corresponds to the LINK[8] element on ANSYS®.
Anterior cruciate, posterior cruciate, lateral and medial collateral ligaments were casted with a bundle of three (3) bar elements for each ligament.
The mean sizes of the cross-sectional areas of the ligaments were taken directly from literature[9] and transcribed on Table 1.
The model described here is composed by femur, tibia, fibula, anterior cruciate ligament, posterior cruciate ligament, medial collateral ligament, and lateral collateral ligament. Tibia and fibula have full restriction to movements, with femur free to all degrees, except for flexion/ extension, whose movement is controlled. Femoral rotation was given as an outlining condition, incremental in time, and the remaining degrees of freedom are free to reach balance. A 100 N follower force was applied to the femur at anteroposterior plane, in order to represent the force applied to femur by the patella. A similar technique was used by Blankevoort and Hhiskes[4] , and by Moglo and Shirazi-Adl[9].

Table 1 – *Cross-sectional areas of main knee ligaments.*

Ligament	Cross-sectional area
Anterior Cruciate Ligament (ACL)	42 mm^2
Posterior Cruciate Ligament (PCL)	60 mm^2
Medial Collateral Ligament (MCL)	18 mm^2
Lateral Collateral Ligament (LCL)	25 mm^2

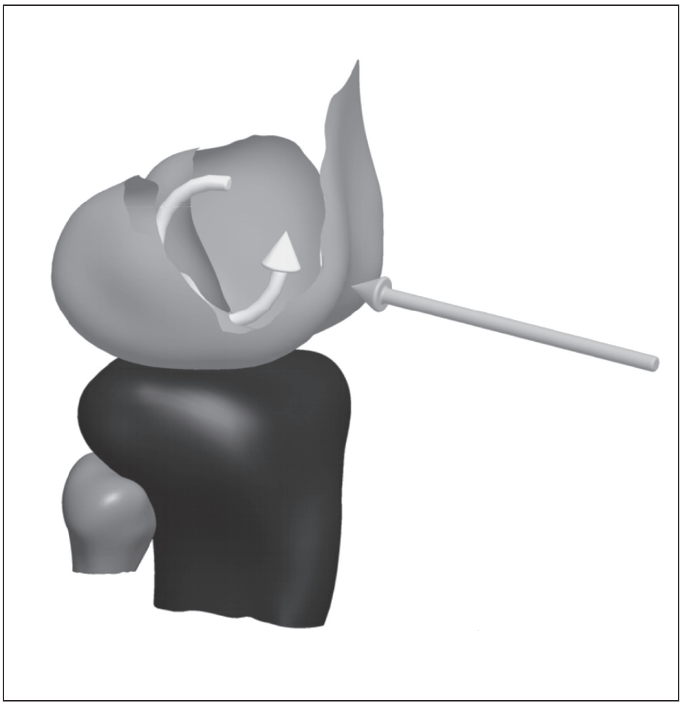

Figure 5 – *External load and imposed movement.*

Human knee has an excellent joint lubrication system. That system is basically formed by the synovial bag and fat cushions, these located in various sites of the joint set. Thus, the friction between joint surfaces is extremely low, and, therefore, was neglected for modeling.

It is also worthy to mention some peculiarities introduced by the contact condition between the bodies. The analysis of issues involving contact between solid materials, or surfaces, usually involves two steps: 1) Search and identification of contact points between surfaces; 2) Introduction of a contact formulation restricting penetrations between solid materials, incorporating normal and tangential efforts (constitutive principle of friction) between surfaces.

Several methods are reported by literature and commercial software of Finite Elements for considering the restrictions imposed by a contact condition. In this study, for the formulation of contact between joint surfaces, we selected the Lagrange's Multipliers method.

On Figure 6, the knee joint system model is depicted, with tibia and fibula on the bottom, and the femur on the top. The geometry presented is already outlined with triangle elements, while ligaments are represented by lines.

RESULTS

We can see on Figure 7 the sequence of movements performed by the femur during a flexion cycle at a medial/ lateral plane.

The results achieved with the model submitted to early pre-load are illustrated on Figures 8 to 14, where values are compared to the correspondent results reported by Mesfar[8], Moglo[9] and Wilson.[10]

We can see on Figure 8 that femoral anteroposterior translation values on tibia are very close to the values reported by Moglol[9] up to approximately 30° of femoral flexion. After this flexion angle value, these get significantly higher. A potential, although not yet proven, explanation for such discrepancy is the absence of menisci in the present model, which were in place in the model designed by Moglo.[9]

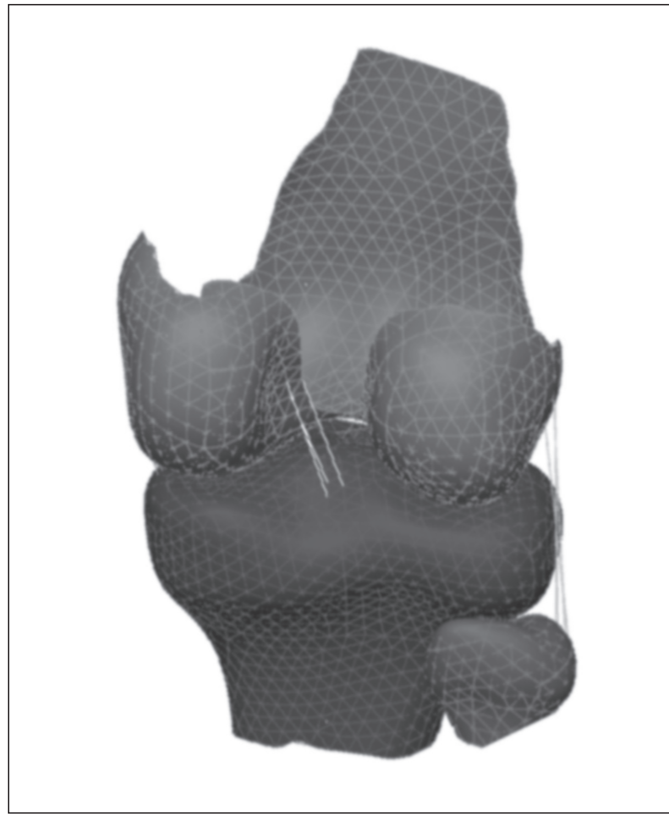

Figure 6 – *Geometrical representation of the model in study.*

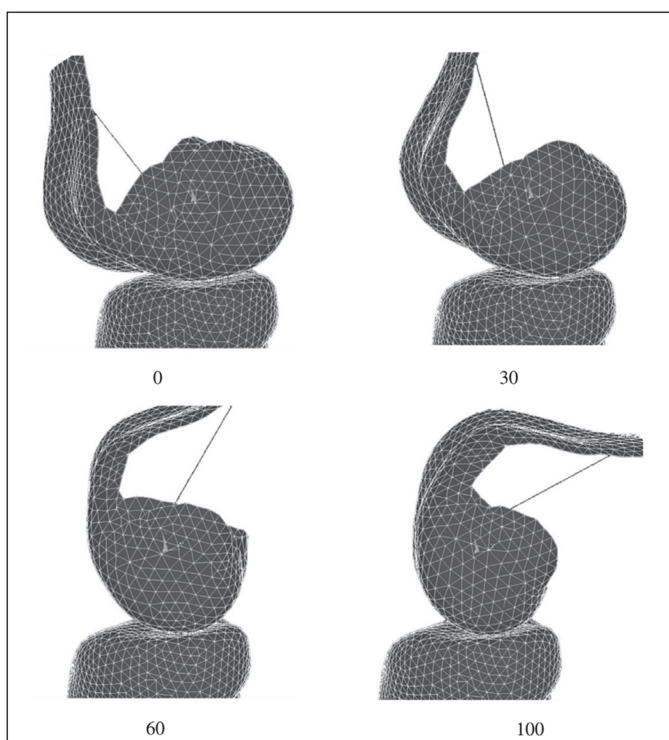

Figure 7 – *Representation of femoral flexion of the model; medial/ lateral view.*

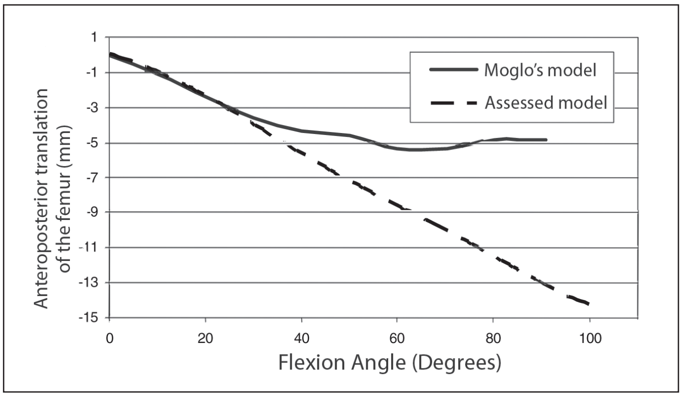

Figure 8 – *Anteroposterior translation of the femur as a function of the flexion angle.*

Figure 9 shows a comparison of tibial abduction/ adduction values in this study with the experimental and numeric values reported by Wilson[10], where the same behavior trend is observed, although with significantly higher values.

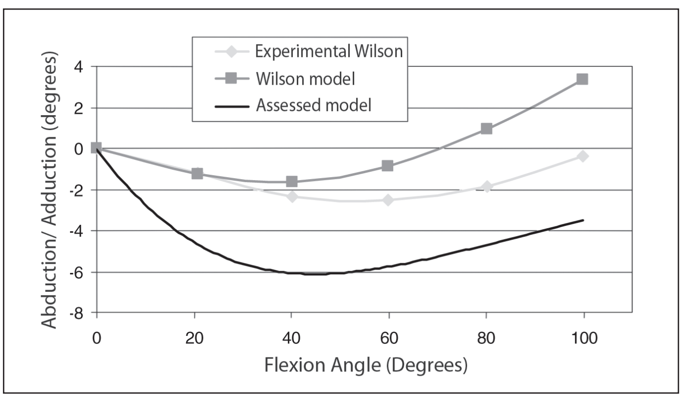

Figure 9 – *Tibial abduction/ adduction as a function of the femoral flexion angle.*

The model was also able to achieve a good representation of the tibial rotation, if compared to the correspondent Wilson[10] values, as presented on Figure 10. However, it presents some variation in these values, characterizing movement instability. This instability might have been caused by the absence of some joint structure with a stabilizing function and also because the method employed here is the one described by Langrange, which assures the non-penetration between surfaces and the absence of friction between joint surfaces, thus, contact surfaces' irregularities directly affect the smoothness of femoral movements.

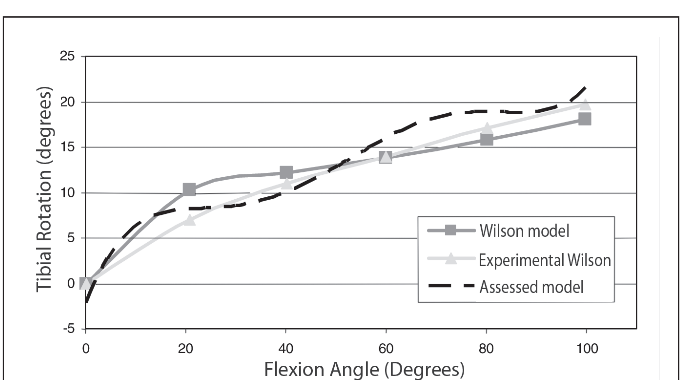

Figure 10 – *Tibial rotation as a function of the femoral flexion angle.*

On Figure 11, the values for force on posterior cruciate ligament obtained by this model are presented and compared to values reported by Moglo[9]. Here, we see that the force values achieved are markedly higher than the ones reported by Moglo[9], but keeping the same behavior trend as a function of the flexion angle. The broader range presented by this model should be associated to an excessive pre-tension applied to the ligament.

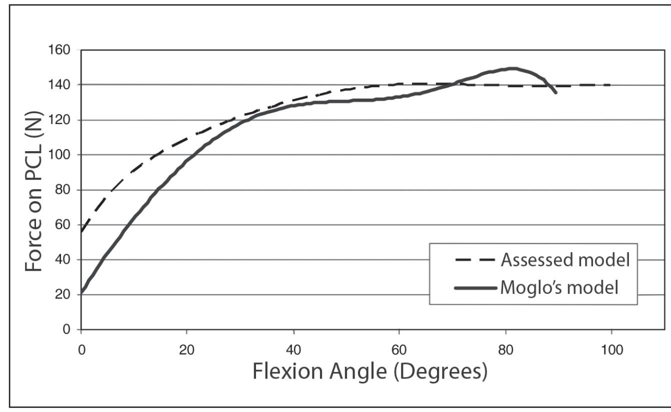

Figure 11 – *Force on PCL as a function of the flexion angle.*

On Figures 12, 13 and 14, forces on LCL, MCL and ACL are respectively shown as a function of the flexion angle. A discrepancy can be seen on Figure 14 for ACL force values obtained here as compared to Mesfar's.[8]

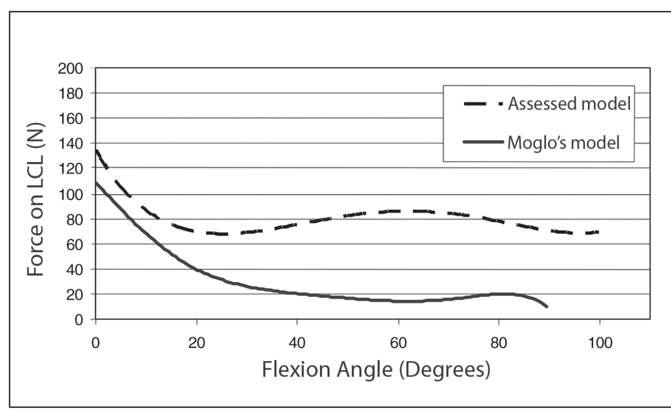

Figure 12 – *Force on lateral collateral ligament as a function of the flexion angle.*

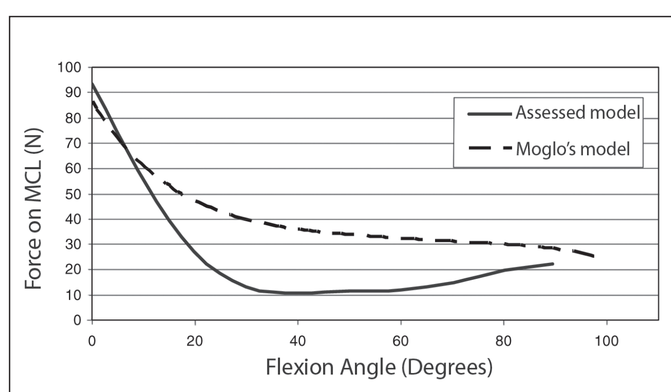

Figure 13 – *Force on medial collateral ligament as a function of the flexion angle.*

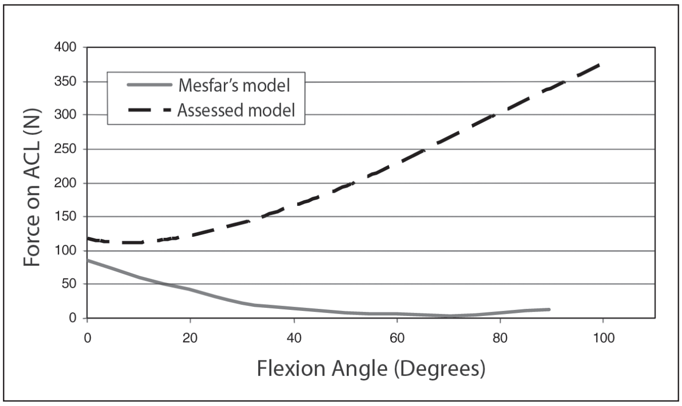

Figure 14 – *Force on anterior cruciate ligament as a function of the flexion angle.*

DISCUSSION AND CONCLUSIONS

A human knee model developed on computer in this study should be understood as a first step in the development of a support tool for medical decision in the area of ligament surgeries and physical therapy. Such model aims to stimulate the biomechanics of a human knee joint, providing data on ligament and contact efforts between joints during knee flexion movement.

From a geometrical point of view, the accuracy of a numeric model to simulate the dynamics and kinematics of a knee joint is strictly associated to a good geometrical representation of joint surfaces, of the numeric model for representing the contact points between these surfaces, of the sites of ligament insertion on the bones, as well as the consideration of all structurally relevant elements and its mechanical properties.

As mentioned in this report, standard geometrical descriptions of the tibia, femur and fibula obtained from the Biomechanics European Laboratory Repository, generated from tree-dimensional reconstitution of plain computed tomography images of cadavers' anatomical pieces have been used. Menisci geometry was not available, and these were ruled out for this primary analysis. Ligaments were remodeled as one-dimensional components, submitted to finite deformations and responding to behavior principles of a non-linear elastic material. Bone pieces were regarded as stiff when compared to ligaments.

The analyses presented on this report focused the efforts experienced by ligaments and femoral rotation and translation movements when a flexion movement was imposed. For providing a qualitative validation between this model and correspondent data obtained from literature, which served as comparison parameters.

Note that above 30° flexion angle, the values and anteroposterior translation of the femur on the tibia become significantly higher. One hypothesis to be corroborated is that this increased anteroposterior translation of the femur on the tibia may be caused by the absence of menisci on this model, which are present on Moglo's model.[9] Similarly, the discrepant force values reported for ACL (Figure 14) are suspected to be associated with the absence of menisci, which are present on Mesfar's model.[8]

Concerning tibial abduction/ adduction, as this is basically governed by the geometry of joint surfaces, the differences on geometrical models used in each study may have produced the difference of values found here and those presented by Wilson.[10]

Even with the differences between models. A good similarity was found for joint movements, but significant differences were seen concerning ligament efforts.

In order to enable a customized (for the patient him/herself) joint set modeling, a three-dimensional geometric model is required for joint pieces, independently from each other, obtained from computed tomography, magnetic resonance, ultra-sound images or by means of other usual medical device, obtaining the precise position of ligaments insertion sites into bones, since these insertion sites can impose bias to the kinematic response of the model. Unfortunately, capturing the geometries of the various knee components from tomography images of an assembled knee (i.e., in vivo) is a complex task, since the required data are obtained from information about density gradients, thus imposing a high level of difficulty in determining the borderline between one piece to another.

By obtaining a geometrical three-dimensional model of the patient him/herself with independent joint pieces, and with the precise ligament insertion site and all structurally relevant elements, the application of the techniques studied here on the customized analysis of surgical interventions on the knee joint region will be possible.[10]

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